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PATENT SPECIFICATION (11) 1 576 588

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(21) Application No. 10298/78 (22) Filed 15 March 1978 (19)
(31) Convention Application No. 783 717 (32) Filed 1 April 1977 in
(33) United States of America (US)
(44) Complete Specification published 8 Oct. 1980
(51) INT. CL.³ H05G 1/62 A61B 6/00 G01N 23/06
(52) Index at acceptance
H4F D12X D18K D18X D27M D30T3 D61 D70P D82A D83B L

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(54) CARDIAC IMAGING SYSTEMS AND METHODS EMPLOYING COMPUTERIZED TOMOGRAPHIC SCANNING

(71) We, OHIO-NUCLEAR, INC., a corporation of the State of Ohio, United States of America, of 29100 Aurora Road, Solon, Ohio 44139, United States of America, do hereby declare the invention, for which we pray that a patent may be granted to us, and the method by which it is to be performed, to be particularly described in and by the following statement:—

This invention relates to cardiac imaging systems and methods employing computerized tomographic scanning.

Computerized tomographic X-ray or gamma ray scanners (CT scanners) reconstruct an image representing a single tomogram of the radiation absorptivity of tissues from data collected from numerous coplanar scan lines. The widest application of CT scanners thus far has been for brain studies. Being stationary when supported in the CT scan circle, all parts of the brain generally remain in the same location during each of the numerous scans required for constructing a single tomographic image. However, involuntary muscular activity makes accurate image reconstruction of other parts of the body difficult. This problem is presented with both basic types of CT scanners, namely, traverse-and-rotate type CT scanners and purely-rotational-type CT scanners.

Heart structures, for example, are in constant motion. While the heart period is on the order of one second, distinct physiological phases of the cardiac cycle, for example, the periods referred to as end systole (ES) and end diastole (ED) have durations of the order of $^1/_{20}$ and $^1/_5$ of a second, respectively. That is, if all of the scan lines needed to reconstruct an image of the heart could be produced in less than $^1/_{20}$ of a second, the motion of the heart would be effectively frozen during either of these periods. This speed, however, is difficult for conventional CT scanners which normally require from about 5 seconds to several minutes to collect the scan data for a single image.

The objectives of cardiac imaging in general are visualizing the sizes of the cardiac chambers, estimating contractilities of the chambers, comparing chamber wall motions, locating aneurisms and areas of myocardial infarction and detecting mitral stenosis. Most of these objectives are, of course, difficult to attain using conventional exposed film X-ray techniques because the differences in absorption or density of heart tissues and blood is not sufficient to confidently distinguish these features at safe radiation dosages and because a tomogram or cross-sectional slice image is not generated.

An electrocardiogram (ECG) is produced by recording the amplitude of electrical activity associated with the heart muscle versus time. In ultrasound imaging, the ECG signal has been used before as a synchronizing device for producing a stop-action image of the heart. See, for example, U.S. Patent No. 3,954,098. Some ultrasound imaging systems have used computerized sorting and assembling multiple images per nominal heart cycle with recorded data from several heart cycles.

Ultrasound imaging differs fundamentally from X-ray imaging. X-rays are not normally reflected detectably by tissue; that portion which is not absorbed is merely transmitted. All conventional X-ray imaging machines operate in the transmission mode. While ultrasound imaging can be carried out in the transmission mode in some instances,

convention ultrasound cardiac imaging, particularly ECG-gated imaging, is only done in the reflecting or echo mode. Ultrasound imaging involves pinpointing each partially reflecting surface for a given pulse of sound energy by measuring the round trip transit time for reception of the echoes, just as in sonar. A single pulse of radiation, however, in the X-ray transmission mode results in a single datum describing the total absorption encountered over the entire path of the X-ray beam; that is, the location of structures is not identifiable from one pulse.

According to a first aspect of the invention there is provided a system for cardiac imaging comprising:

computerized tomographic scanner means for generating and storing image data indicative of a patient's tissue density and for reconstructing a tomographic image from said image data, said scanner means including means for generating at least one beam of radiation and scan means for scanning the beam of radiation relative to the heart;

cardiac cycle monitoring means for producing a repeating pulse signal indicative of the same functional point in each successive cardiac cycle of the patient; and

control means responsive to said pulse signal for causing said scanner means to store image data in the same selectable phase in successive cardiac cycles for use in constructing a tomographic image of the heart in said phase,

said control means including selectable delay means for delaying said pulse signal for a selectable duration thereby selecting said selectable phase in the cardiac cycle.

According to a second aspect of the invention there is provided a method of producing at least one tomographic image of at least one cross-section of a patient's heart in a selectable phase of the cardiac cycle with a computerized tomographic scanner, the method comprising:

sensing the same functional point in each successive cardiac cycle of the patient and producing a pulse signal indicative of said same functional point;

delaying said pulse signal for a selectable duration for selecting the phase of the cardiac cycle to be imaged;

generating at least one beam of radiation;

scanning said beam relative to the patient's heart;

generating image data indicative of the radiation attenuation along a plurality of successive scan lines intersecting the patient's heart;

controlling the storage of image data with said delayed signal for storing the image data during the same selectable phase in successive cariac cycles; and

constructing said tomographic image of the heart in said selectable phase.

Embodiments of the invention described below produce an image of the radiation attenuation of the heart at a desired phase of the cardiac cycle. The patient's ECG signal is, in one embodiment, employed in a traverse-and-rotate-type CT scanner as a time base for triggering the beginning of a traverse such that the travelling beam reaches the heart at a desired phase of the cardiac cycle. In another embodiment, comprising a purely-rotational-type CT scanner, continuously generated scan data is only stored for corresponding phases of successive cardiac cycles. Alternatively, gating of the beams themselves by shuttering or switching the power supply can be controlled by the ECG signal. A pacemaker is used to stabilize the cardiac period. Also used is a system for recognizing unacceptable variations in the cardiac period and discarding corresponding scan data. In a traverse-and-totate-type fan-beam CT scanner, the effective beam width is narrowed to reduce the duration of the traverse of the heart.

The invention will now be further described, by way of illustrative and non-limiting example, with reference to the accompanying drawings, in which:

Figure 1 is an ECG waveform;

Figure 2 is a block diagram illustrating a system embodying the invention and incorporating an ECG-controlled CT scanner;

Figure 3 is a schematic diagram indicating a CT scanner beam's traverse in relation to the patient's heart;

Figure 4 is a schematic diagram of a traverse-and-rotate-type fan-beam CT scanner illustrating the beam pattern intersecting the patient's heart;

Figure 5 is a schematic representation of the beam pattern of a purely-rotational-type CT scanner; and

Figures 6, 7 and 8 are detail block diagrams illustrating different systems of ECG-gating for a purely-rotational-type CT scanner.

The ECG waveform shown in Figure 1 represents features designated by the letters P, Q, R, S, and T. The group of features Q, R, and S is referred to as the QRS complex, in which the R-feature or R-wave is the most prominent, highest amplitude feature of the entire ECG. Moreover, the narrow pulse width of the QRS complex, and in particular the R-wave, provides a digital clock pulse for timing the cardiac cycle.

The cardiac cycle is usually defined as beginning with the R-wave and continuing until the occurrence of the next R-wave. Heart functions are characterized by two distinct periods called systole and diastole. In systole, the heart muscle is contracting the volume of the left ventricle to pump the contents out through the aortic valve. During diastole, the left ventricle is filling

through the mitral valve. At the end of systole (ES), the left ventricle has its smallest volume since it has contracted to pump blood out. The end of diastole (ED) is the point at which the left ventricle has its largest volume since it is filled with blood ready to be pumped out. These two extremes of heart function, end of systole and end of diastole, are of interest, for example, in determining fractional ejection, i.e., the ratio of minimum-to-maximum ventricular volume. Each of these features, end of systole and end of diastole, lasts for an interval of the order of $^1/_{10}$ second and occurs once every cardiac cycle.

Fig. 2 shows a conventional traverse-and-rotate-type CT scanner having a scan circle 12 defining a scan plane in which the patient 14 is positioned such that the scan plane preferably intersects the left ventricle, left atrium or aortic root of the heart. The mechanical operation of the traverse-and-rotate mechanism and the beam shutter is controlled by a scanner controller 16. An external pacemaker 18 may be employed to stabilize the cardiac period of a patient with irregular heart rate. The ECG signal from the patient is applied by an isolation amplifier 20 to a QRS complex detector 22 whose output is a digital timing pulse corresponding to each R-way of the patient's live ECG signal. The items 20 and 22 are commercially available units, for example, Hewlett Packard Corporation, Model Nos. 7807C and 7830A. The output of the QRS detector 22 is fed to a delay timing circuit 24 which provides a trigger pulse to the CT scanner controller 16.

If the objective is to acquire an image of the heart at the end of diastole with the left ventricle fully expanded, the trigger pulse is timed to be applied to the scanner controller 16 sufficiently in advance of the end of diastole so that the scanner controller 16 can begin the traverse of the radiation beam, as shown in Fig. 3, from the point x_0 so that by the time the beam travels distance D_1 to position x_1 where it first begins to intersect the heart, the heart will be in the end of diastole stage. Of course, the beam must be traversing at a rate sufficient to traverse the width of the heart in this presentation in approximately less than $^1/_{10}$ of a second. Thus, the distance D_2 from point x_1 to the point x_2 would determine the minimally acceptable speed of traverse. The consequence of travelling too slowly through the distance D_2 in Fig. 3 would be a blurring of some of the moving heart structures.

The delay time provided by the circuit 24 in Fig. 2 is determined by three parameters: (1) the speed of the traverse of the beam; (2) the position of the heart in the scan circle 12; and (3) a prediction of when the particular phase of interest, for example end of diastole, will occur in the average or nominal cardiac cycle of the patient. The speed of traverse of the beam is normally a known constant value. However, the speed can be monitored during the scan to compensate in successive scans for any variations in the average scan speed. The position of the heart can be determined in two ways: the heart's position can be considered by adjusting the patient's position in the scan circle or the patient can be prescanned and the location of the heart determined by the scanner operator from the reconstructed image. Prediction of the time that the heart will be in a particular phase, such as end diastole, requires a knowledge of the heart rate, as measured by the interval between R-waves, and the average time elapsed from an R-wave up to the phase of interest. This information is derived from the patient's electrocardiogram. It could also be derived from a phonocardiogram or pressure measurements.

These parameters are taken into consideration in setting the delay implemented by the timing circuit 24. After the R-wave signal from the QRS detector 22, the delay timing circuit 24 pauses before issuing a trigger pulse for an interval of time which can be represented as follows:

$$T_{ED} - D_1/R_{avg}$$

where T_{ED} is the predicted time from a given R-wave to the beginning of the end of the diastole phase; D_1 is the position of the heart in terms of the distance the beam covers from the starting point x_0 until reaching the centre wall of the heart (or some other point of interest); R_{avg} is the average speed of the traversing beam; and D_1/R_{avg} is the predicted elapsed time from the beginning of the traverse to the point where the centre beam intersects the centre of the heart.

For any patient the period of the cardiac cycle, as shown in Fig. 1, from one R-wave to the next R-wave always varies to some degree. The timing of the triggering of a traverse is based solely on the occurrence of the last R-wave and the predicted time for beginning of the end of diastole or end of systole whichever phase is being imaged. It is entirely possible that the prediction may not be borne out. If the cardiac cycle, which the scanner is preparing to sample, is one of sufficiently increased or reduced period, the end of diastole or end of systole will occur at a significantly different time. Thus, it is advisable to place tolerance limits on the cardiac period in order to distinguish acceptable and unacceptable scan data.

The system of Fig. 2 merely illustrates one form of digital circuitry for performing heart period dissemination. In practice, it may be preferable to implement these func-

tions with software using the CT scanner computer associated with image processing the machine control. The output of the QRS detector 22 is passed to the reset input of a digital counter 26 clocked, for example, at one hundred or one thousand Hertz by a stable frequency oscillator 28. The parallel binary output of the counter 26 is passed via a latch circuit 30 to a subtractor circuit 32. The latch 30 operates as a digital sample-and-hold circuit which holds the count attained by the counter 26 immediately before being reset by the next R-wave. The number contained in latch 30 is compared by the subtractor 32 to the number held in storage 34 representing the nominal period of the patient's cardiac cycle. The difference between the counts for the actual and nominal periods is passed to a comparator 36. A reference number indicating a tolerance limit on the difference between the actual and nominal periods is provided by the limit circuit 38. If the difference exceeds the limit provided by the circuit 38, the binary comparator output alerts an image processing unit 40 associated with the CT scanner 10 to discard the scan data corresponding to the irregular period.

The actual tolerance limits on cardiac period depend on the phase being imaged. For example, the tolerance for imaging end of diastole will be smaller than the tolerance for end of systole since the interval to end of diastole is generally regarded as proportional to the cardiac period and comes at the very end of the cardiac cycle. Thus, in addition to setting the delay timing in accordance with the phase of interest, the tolerance limits for an acceptable cardiac period should also be adjusted accordingly.

Instead of employing a single beam as shown schematically in Fig. 3, several CT scanning devices currently on the market, such as the Delta-Scan model marketed by Ohio-Nuclear, Inc., traverse with a fan-shaped pattern of beams as shown in Fig. 4. The fan-beam pattern covers a width W_1 where it intersects the heart. The consequence of the width of the beam pattern is that it takes longer for the entire plurality of beams to traverse the heart from point x_1 to point x_2. As is the case in Fig. 4, if the width of the beam is approximately equal to D_2, i.e. the width of the heart in the plane in the scan direction, the time relative to a narrow beam scan will be doubled for a full heart traverse of all of the beams in the fan pattern. To alleviate this problem, the effective beam width can be narrowed, for example to W_2, by ignoring or "throwing out" data from several of the peripheral detectors. For example, the data from the two outermost detectors on either side can be ignored. Alternatively, a shutter can be employed as shown in Fig. 4 to block certain peripheral rays thus narrowing the actual pattern. The effect of either of these remedies is to narrow the fan-beam width at the heart so that the distance D_2 can be traversed more quickly. The faster the interval D_2 is traversed, the less motion will be present to cause blurring in the image. Thus, the precision of the stop-action effect can be increased by omitting data from peripheral beams to achieve a shorter effective time window. Blocking the beams instead of ignoring the data from the detectors has the advantage of eliminating unnecessary X-ray dosage. However, removing several of the beams will cause a slight increase in the overall scan time for completing the image.

Cardiac imaging can also be accomplished with a purely rotational CT scanner as shown in Fig. 5 wherein α is the rotational axis. In this case, however, since rotation of the source and detectors is continuous, instead of triggering the mechanical traverse at the right point so that data is acquired in the phase of interest, scan data is generated continuously. The patient's ECG signal can be used as in Fig. 6 to gate the storage of data by the image processing unit 40. The delay timing circuit is thus used to open a time window during the appropriate phase in each cardiac cycle in which data is collected. Several phases can be gated with the ECG signal or even a full set of images covering every distinct physiological point in the cardiac cycle can be generated. These images can be sequenced then to produce a movie or cine presentation of the subject's heart. This alternative also includes the possibility of a semi-circle of stationary detectors and a rotating source, or stationary source(s) and rotating detectors.

In order to minimize X-ray dosage to the patient, the X-ray tube could be gated on and off as indicated by the gating of X-ray power supply 42 in Fig. 7, or dynamically shuttered as indicated in Fig. 8 by the gating of beam shutter control 44, which would open the shutter blocking the beams only during the physiological phase of interest during each cardiac cycle.

In addition to cardiac gating, chest motion from breathing can be removed from the image by using pulmonary gating. Scan data would only be stored at times when the phase of the cardiac cycle under investigation coincided with a particular phase of the pulmonary cycle.

The above-described embodiments are intended to be illustrative, not restrictive. For example, the use of the interval end of diastole or end of systole is intended to be illustrative of the use of any interval of interest. Moreover, the same techniques disclosed herein may be applicable to other physiological functions besides the heart. The invention is applicable, of course, to

any type of beam transmission subject to differential tissue absorption such as X-rays, γ rays, etc.

WHAT WE CLAIM IS:—

1. A system for cardiac imaging, comprising:

computerized tomographic scanner means for generating and storing image data indicative of a patient's tissue density and for reconstructing a tomographic image from said image data, said scanner means including means for generating at least one beam of radiation and scan means for scanning the beam of radiation relative to the heart;

cardiac cycle monitoring means for producing a repeating pulse signal indicative of the same functional point in each successive cardiac cycle of the patient; and

control means responsive to said pulse signal for causing said scanner means to store image data in the same selectable phase in successive cardiac cycles for use in constructing a tomographic image of the heart in said phase,

said control means including selectable delay means for delaying said pulse signal for a selectable duration thereby selecting said selectable phase in the cardiac cycle.

2. A system according to claim 1, comprising irregular cardiac cycle sensing means comprising discriminator means responsive to said pulse signal for producing an output signal to command said scanner means to discard image data when the corresponding cardiac period varies from the patient's nominal cardiac period beyond predetermined limits whereby data from irregular cardiac cycles is prevented from degrading the tomographic image.

3. A system according to claim 2, wherein said discriminator means comprises means for adjusting said limits according to the selected phase in the cardiac cycle.

4. A system according to claim 1, claim 2 or claim 3, comprising external pacemaker means for stabilising the patient's cardiac period.

5. A system according to any one of the preceding claims, wherein said scanner means comprises a traverse-and-rotate-type scanner arrangement and said control means includes means responsive to said pulse signal for generating a trigger pulse to initiate a beam traverse such that the travelling beam will reach the heart at the time of the selected phase in the cardiac cycle.

6. A system according to claim 5, wherein said traverse-and-rotate type scanner arrangement includes means for generating a plurality of co-planar beams of radiation in a fan-shaped pattern, the system being non-responsive to image data associated with at least one peripheral beam in said fan-shaped pattern, or being arranged to prevent the generation of image data associated with such beam, in order to narrow the effective width of said pattern of beams.

7. A system according to claim 6, which includes means for shielding the patient from at least one said peripheral beam to prevent the generation of any image data associated therewith.

8. A system according to any one of the preceding claims, comprising means for producing a pulmonary pulse signal indicative of the same functional point in each successive pulmonary cycle of the patient, said control means further being responsive to said pulmonary pulse signal for causing said scanner means to store image data in the same selectable phase in successive cardiac cycles and the same particular phase of the pulmonary cycle in order to eliminate errors from chest motion due to breathing.

9. A system according to any one of the preceding claims, wherein said control means is operative to cause said scanner means to generate image data only in at least one selectable phase of the cardiac cycle.

10. A system according to any one of claims 1 to 4, wherein said scanner means is a purely-rotational-type scanner system, and said control means includes means responsive to said pulse signal for generating a gate signal having a timing and duration corresponding to the predicted occurrence of said selectable phase cycle and means responsive to said gate signal for enabling image date storage during said gate signal.

11. A system according to any one of claims 1 to 4, wherein said scanner means is a purely-rotational-type CT scanner system, and said control means includes means responsive to said pulse signal for generating a gate signal having a timing and duration corresponding to the predicted occurrences of said selected phase, and means responsive to said gate signal for gating the beam of radiation on and off synchronously with said selectable phase.

12. A system according to claim 11, wherein said means for gating the beam includes controllable shutter means for blocking the beam.

13. A method of producing at least one tomographic image of at least one cross-section of a patient's heart in a selectable phase of the cardiac cycle with a computerized tomographic scanner, the method comprising:

sensing the same functional point in each successive cardiac cycle of the patient and producing a pulse signal indicative of said same functional point;

delaying said pulse signal for a selectable duration for selecting the phase of the cardiac cycle to be imaged;

generating at least one beam of radiation;

scanning said beam relative to the patient's heart;

generating image data indicative of the radiation attenuation along a plurality of successive scan lines intersecting the patient's heart;

controlling the storage of image data with said delayed signal for storing the image data during the same selectable phase in successive cardiac cycles; and

constructing said tomographic image of the heart in said selectable phase.

14. A method according to claim 13 wherein said scanning step includes:

linearly traversing said beam relative to the heart, commencing said linear traversing at said same functional point in each successive cardiac cycle whereby the traversing beam reaches the heart as the time of said selectable phase in the cardiac cycle; and

rotating said beam after each linear traverse.

15. A system for cardiac imaging, the system being substantially as herein described with reference to Figures 1, 2 and 3 or any one of Figures 4 to 8 of the accompanying drawings.

16. A method of producing at least one tomographic image of at least one cross-section of a patient's heart in a selectable phase of the cardiac cycle with a computerized tomographic scanner, the method being substantially as herein described with reference to Figures 1, 2 and 3 or any one of Figures 4 to 8 of the accompanying drawings.

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Printed for Her Majesty's Stationery Office by Burgess & Son (Abingdon), Ltd.—1980.
Published at The Patent Office, 25 Southampton Buildings, London, WC2A 1AY, from which copies may be obtained.

FIG. 1

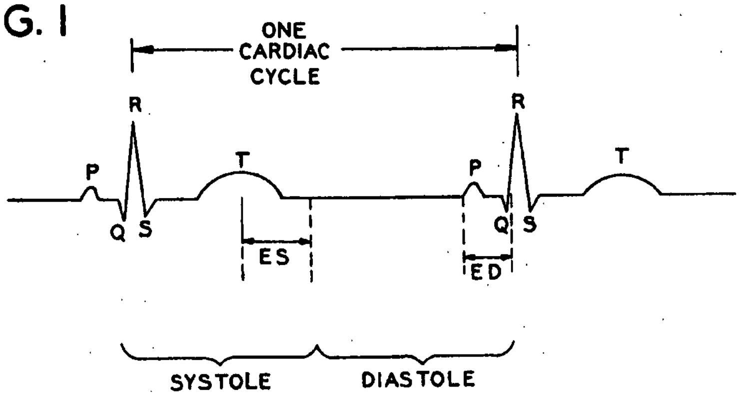
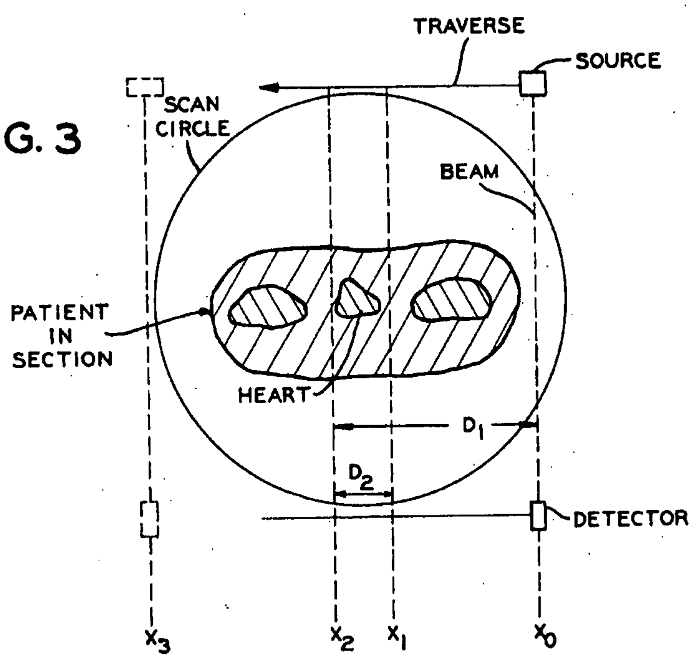

ONE CARDIAC CYCLE
R
P
T
Q
S
ES
ED
SYSTOLE
DIASTOLE

FIG. 3


TRAVERSE
SOURCE
SCAN CIRCLE
BEAM
PATIENT IN SECTION
HEART
D1
D2
DETECTOR
X3
X2
X1
X0

1576588 COMPLETE SPECIFICATION

4 SHEETS *This drawing is a reproduction of the Original on a reduced scale*

Sheet 2

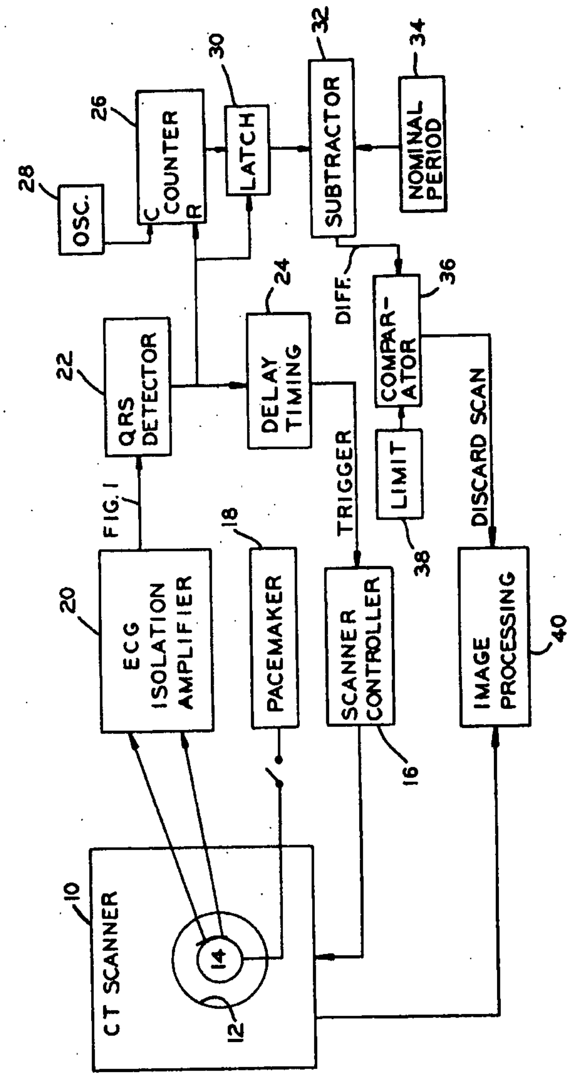
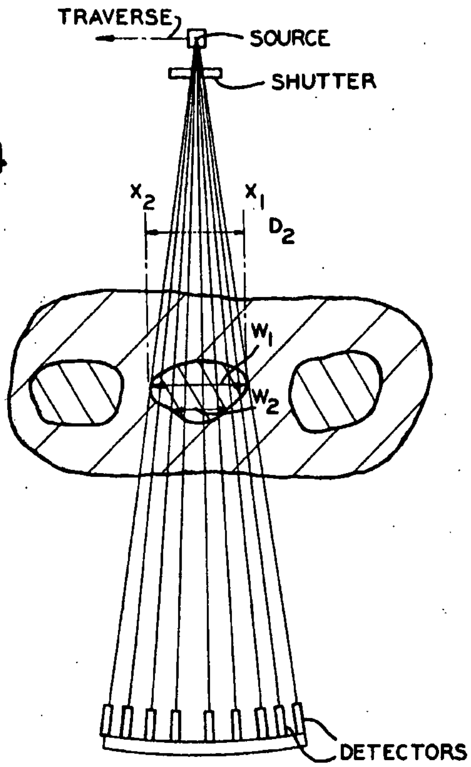


FIG. 2

FIG. 4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x_1
D_2
W_1
W_2
DETECTORS

FIG. 5

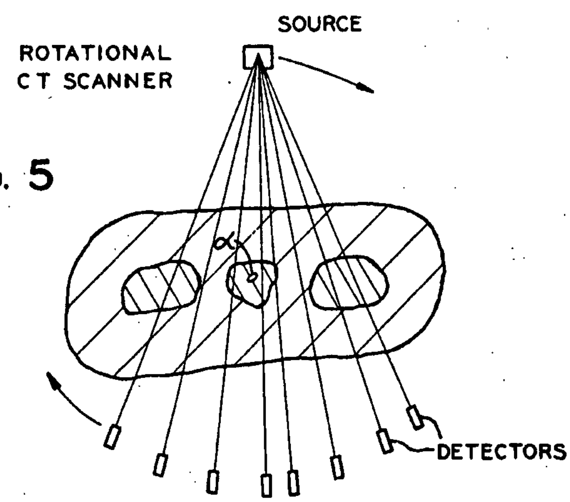
SOURCE
ROTATIONAL
CT SCANNER
α
DETECTORS

FIG. 6

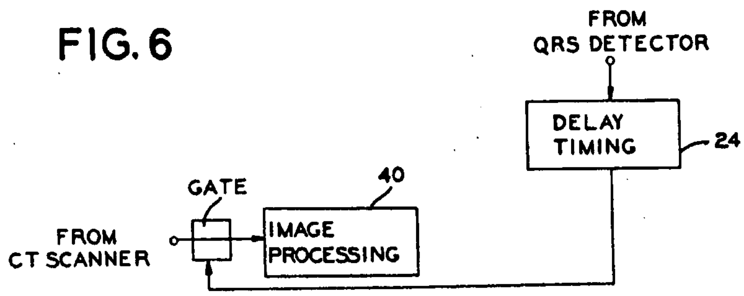

FIG. 7

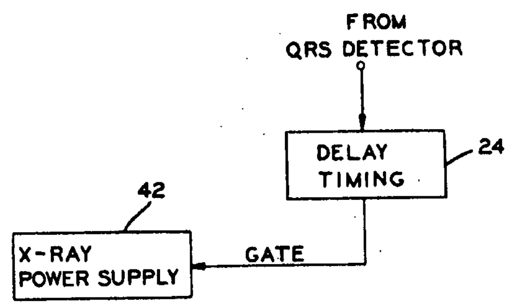

FIG. 8